A validation study of a new instrument for low cost bite force measurement

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ABSTRACT

Quantitative assessment of force in masticatory muscles is not a routine clinical test, probably due to the lack of an “easy-to-use” device. Aim of this study is 1) to present a low cost bite force instrument located in a custom-made housing, designed to guarantee a comfortable and effective bite action, 2) to evaluate its mechanical characteristics, in order to implement it in clinical settings and in experimental setups.

Linearity, repeatability and adaptation over time were assessed on a set of four different sensors in bare and housed condition. Application of the housing to the transducer may appreciably alter the transducer’s response. Calibration of the housed transducer is thus necessary in order to correctly record real bite force. This solution may represent a low cost and reliable option for biting force measurement and objective assessment of individual force control in the scientific and clinical setting.

protrusion of the mandible, which affects the biting force, while placement of the sensor in the premolar or molar region makes the precise repositioning of the sensors more difficult, thus affecting repeatability;

3) costs: extra oral apparatuses have been developed, connected with servo controlled motors, adequate to investigate motor function and reflexes (Turker et al., 2004, van der Bilt et al., 2006) however, due to their cost and complexity, these systems are more suited for research purposes than for routine clinical examination. Some authors have used load cells mounted on a customized dental (Castroflorio et al., 2008) device to evaluate bite force, but also this approach results in a complex and costly technical procedure.

Aim of this study is to present a much simpler and practical solution, based on a low cost and versatile piezo-resistive force sensor, adequate to measure bite force in a clinical setting. A thin commercial transducer is accommodated within a protective rubberized housing and characterized from the electro-mechanical point of view.

2. MATERIAL AND METHODS

2.1 Sensor housing description

Force measurement was based on the piezoresistive force transducer Flexiforce A201 (Tekscan, Boston, MA, USA), featuring a load range of 100 lb, equivalent to 440 N, and a sensitivity of 0.01 V/N. The *FlexiForce* force sensor is a flexible printed circuit with at one end an active sensing area made of pressure-sensitive ink of 1 cm of diameter. The circuit is embedded within two polyester film layers with a final thickness of 0.2 mm. (Fig.1A)

Forces exerted on the active sensing area cause a roughly proportional change of the sensor's conductance. A special housing was developed in order to protect the sensor from mechanical damage and to reduce discomfort for the subject during clenching. The force transducer was inserted in a home made "sandwich structure" composed of different plastic foils and a steel disc, stuck together by bi-adhesive film. The multi-layer "L" shaped housing was developed as shown in Fig. 1B and Fig. 1C. The external layer is made of a silicone rubber material, commonly used in preparing dental orthotics (Bioplast – Scheudental - Germany). This layer provides the possibility of small yielding of the surface under the teeth, thereby generating a wider contact surface, thus lowering local pressure. It also provides improved comfort during clenching, as the active area of the flexiforce transducer and was measured by a load cell. In order to assess relative and absolute reliability of the measurement in both conditions, this procedure was performed on the bare and on the housed sensor and repeated after 24 hours.

In addition, in order to test the dependence of the sensor response on the contact surface, four transducer were loaded (300 N) against a smooth or rough Plexiglas surface.

Finally, the application of a constant load of 390 N for 4 min was performed to investigate the output drift over time.

Electric signals generated by the cell and by the Flexiforce transducer were amplified, sampled at 15Hz and digitally-converted (16 bit) by a dedicated hardware (Cal4met, OT Bioelettronica, Torino, Italy) and transmitted to a personal computer (via USB). Numerical data were acquired and stored by a custom-made software written in Matlab (Mathworks, Natick, Massachusetts, USA). Measures from the load cell are expressed in Newton (N), while readings from the sensors are in Volt (V). Linear fitting of sensor load/unload-response curve provided a measure of the sensor sensitivity (slope) and of linearity (R-squared) in both the bare and housed condition. Hysteresis has been defined as difference between the areas under the loading and unloading curves, divided to the area under the loading curve, and the relative means were calculated on absolute values. Relative reliability was assessed by Intraclass Correlation Coefficient (ICC) and absolute reliability by Standard Error of Measurement (SEM) and by Bland and Altman's 95% Limits of Agreement (LOA). Calculations were conducted using "R" software (R Core Team, 2013).

3. RESULTS

3.1 Sensor characterization

The characteristic load-response curves of four different sensors in bare and housed conditions have been measured in two consecutive days. Their qualitative behaviour is given in Fig. 2.

The different sensors showed a similar load-response curve, slightly decreasing the slope at increasing load. Regression analysis of the curve for each sensor in bare and housed condition and in the two days was performed. (Tab. 1). It can be observed that the slope of the regression line, i.e. the sensor sensitivity, may appreciably change in the different sensors (e.g.: 0.004-0.005 V/N, bare sensors, day 1). Moreover, it was systematically higher in the housed than in the bare condition. As for the intercept, ranging between 0.15

surface the force reading was shown to decrease by as much as 40%.

The observed non linearity is very small (r^2 >0.94) and is most likely to be attributed to the signal conditioning, based on a non inverting amplifier and also implementing low-pass filtering and electrical isolation. This non-linearity and the individual variability of the sensitivity may potentially result in a systematic error. However this error may be easily prevented by implementing a multi-point sensor calibration, in the specific load range of the measurement. In this way reliable recording of absolute force values can be achieved. Due to the large depedence of the sensor response on the contact surfaces, it is advisable to apply the housing prior to calibration.

As already mentioned in the results, a little offset (sensor output at 0 load) is introduced by the housing. This offset needs to be accounted for by the calibration procedure and removed, in order to prevent large errors at low load conditions.

Although we did not assessed the effect of temperature change on the sensor output, the variation reported by the manufacturer in the user manual of the flexiforce sensor (https://www.tekscan.com/support/faqs/flexiforce-user-manual) is about 0.36%/°C. If we account for few degrees of temperature difference between the sensor when tested at the bench and when located in the mouth, this variation would be in the order of 3-4%. However, a similar variation is expected to occur in all subjects thus producing negligible changes in the comparisons between subjects or between sides of the same subject.

Sensors operating in the mouth are also potentially exposed to high humidity, whose effect has not been tested in the present study, however exposure to saliva and humidity may be prevented by shielding the sensor with a latex or nitrile finger glove.

Plastic and rubber shielding of the sensor was proven to improve comfort and stability of the force signal (Fernandes et al., 2003, Waltimo and Kononen, 1993). The present results indicate that, although increasing the sensitivity of the sensor, as discussed above, the housing does not affect its linearity. The nominal working range of the sensor (440 N) is adequate for use with adult subjects. In case a wider or smaller range of registration is needed, it could be scaled up or down by adjusting the sensor's dynamic range by tweaking its external drive circuit. The values of interday relative and absolute reliability obtained by testing the sensors in bare and housed condition were quite high. Bland and Altman plots revealed a slightly system uses a film sensor to record the distribution of occlusal forces which is graphically described over the occlusal surface by a qualitative colorimetric representation of dental contactintensity. The information given by the system can support clinical decision in prosthetic dentistry, but has limited value in giving information about the neuromuscular control of the bite force.

In other studies film sensors were embedded in splints for long duration recordings (Baba et al., 2003, Takeuchi et al., 2001). Recently Castroflorio et al. (2008) proposed a system of bite force registration that is based on customized upper and lower dental appliances. These systems must be customized for each subject with significant extra costs. In addition, the complexity of the procedure limits its application within dentistry departments. The piezo-resistive force transducer used in the present study has comparable shape and dimensions to the one used by Fernandes et al. (2003) and shows similar characteristics in terms of measurement errors, making this sensor a potentially useful tool for routine clinical examinations, provided sensor calibration in its working condition is carefully accomplished.

5. CONCLUSIONS

A simple instrument to measure bite force, based on a commercially available sensor inserted in a protective housing and connected to a simple hardware has been developed. The limited thickness of the developed sensor, the easy repositioning of the device in the mouth and its low cost overcome the most common problems encountered in bite force measuring. Indeed, the housing of the sensor can be assembled by hand and the signal conditioning can be obtained by adopting the hardware recommended by the manufacturer or another equivalent solution like the one we have used in the present work.

The characterization tests showed that the presence of the housing layers does not worsen, but on the contrary enhance the sensitivity of the sensor. Accurate preconditioning and subsequent calibration of the housed sensor in its working conditions, by acquisition of the I/O response curve over the load range of interest, is recommended in order to account for the many limitation including non linearity, individual variability of the sensitivity and dependence of the output response on the nature of the contact surfaces.

The proposed transducer may be a handy solution to assess the bite force in the clinical setting and a valid support for standardization of EMG studies on jaw-closing muscles.

Miura H, Watanabe S, Isogai E, Miura K. Comparison of maximum bite force and dentate status between healthy and frail elderly persons. Journal of Oral Rehabilitation. 2001;28:592-5.

Ortug G. A new device for measuring mastication force (Gnathodynamometer). Ann Anat. 2002;184:393-6.

Osborn JW, The temporomandibular ligament and the articular eminence as constraints during jaw opening. Journal of Oral Rehabilitation. 1989;16:323-333

Pereira LJ, Steenks MH, de Wijer A, Speksnijder CM, van der Bilt A. Masticatory function in subacute TMD patients before and after treatment. Journal of Oral Rehabilitation. 2009;36:391-402.

R Core Team. R: A language and environment for statistical computing. Vienna, Austria: R Foundation for Statistical Computing; 2013.

Rismanchian M, Bajoghli F, Mostajeran Z, Fazel A, Eshkevari P. Effect of implants on maximum bite force in edentulous patients. Journal of Oral Implantology. 2009;35:196-200.

Rottner K, Richter EJ. Effect of occlusal morphology on the accuracy of bite force measurements using thin film transducers. International Journal of Prosthodontics. 2004;17:518-23.

Schimmel M, Leemann B, Christou P, Schnider A, Muller F, Kiliaridis S. Masseter muscle thickness in hospitalised stroke patients. Journal of Oral Rehabilitation. 2010;37:726-30.

Schramm A, Classen J, Reiners K, Naumann M. Characteristics of sensory trick-like manoeuvres in jaw-opening dystonia. Movement Disorders. 2007;22:430-3.

Serra CM, Manns AE. Bite force measurements with hard and soft bite surfaces. Journal of Oral Rehabilitation. 2013;40:563-8.

Takeuchi H, Ikeda T, Clark GT. A piezoelectric film-based intrasplint detection method for bruxism. Journal of Prosthetic Dentistry. 2001;86:195-202.

Testa M, Geri T, Gizzi L, Petzke F, Falla D. Alterations in Masticatory Muscle Activation in People with Persistent Neck Pain Despite the Absence of Orofacial Pain or Temporomandibular Disorders. J Oral Facial Pain Headache. 2015a;29:340-8.

Testa M, Geri T, Signori A, Roatta S. Visual Feedback of Bilateral Bite Force to Assess Motor Control of the Mandible in Isometric Condition. Motor Control. 2015b;19:312-24.

Testa M, Rolando M, Roatta S. Control of jaw-clenching forces in dentate subjects. Journal of Orofacial Pain. 2011;25:250-60.

Trawitzki LV, Silva JB, Regalo SC, Mello-Filho FV. Effect of class II and class III dentofacial deformities under orthodontic treatment on maximal isometric bite force. Archives of Oral Biology. 2011;56:972-6.

Turker KS, Brinkworth RS, Abolfathi P, Linke IR, Nazeran H. A device for investigating neuromuscular control in the human masticatory system. Journal of Neuroscience Methods. 2004;136:141-9.

van der Bilt A, Engelen L, Pereira LJ, van der Glas HW, Abbink JH. Oral physiology and mastication. Physiology and Behavior. 2006;89:22-7.

Waltimo A, Kononen M. A novel bite force recorder and maximal isometric bite force values for healthy young adults. Scandinavian Journal of Dental Research. 1993;101:171-5.

Wang MQ, He JJ, Zhang JH, Wang K, Svensson P, Widmalm SE. SEMG activity of jaw-closing muscles during biting with different unilateral occlusal supports. Journal of Oral Rehabilitation. 2010;37:719-25.

Figure 2
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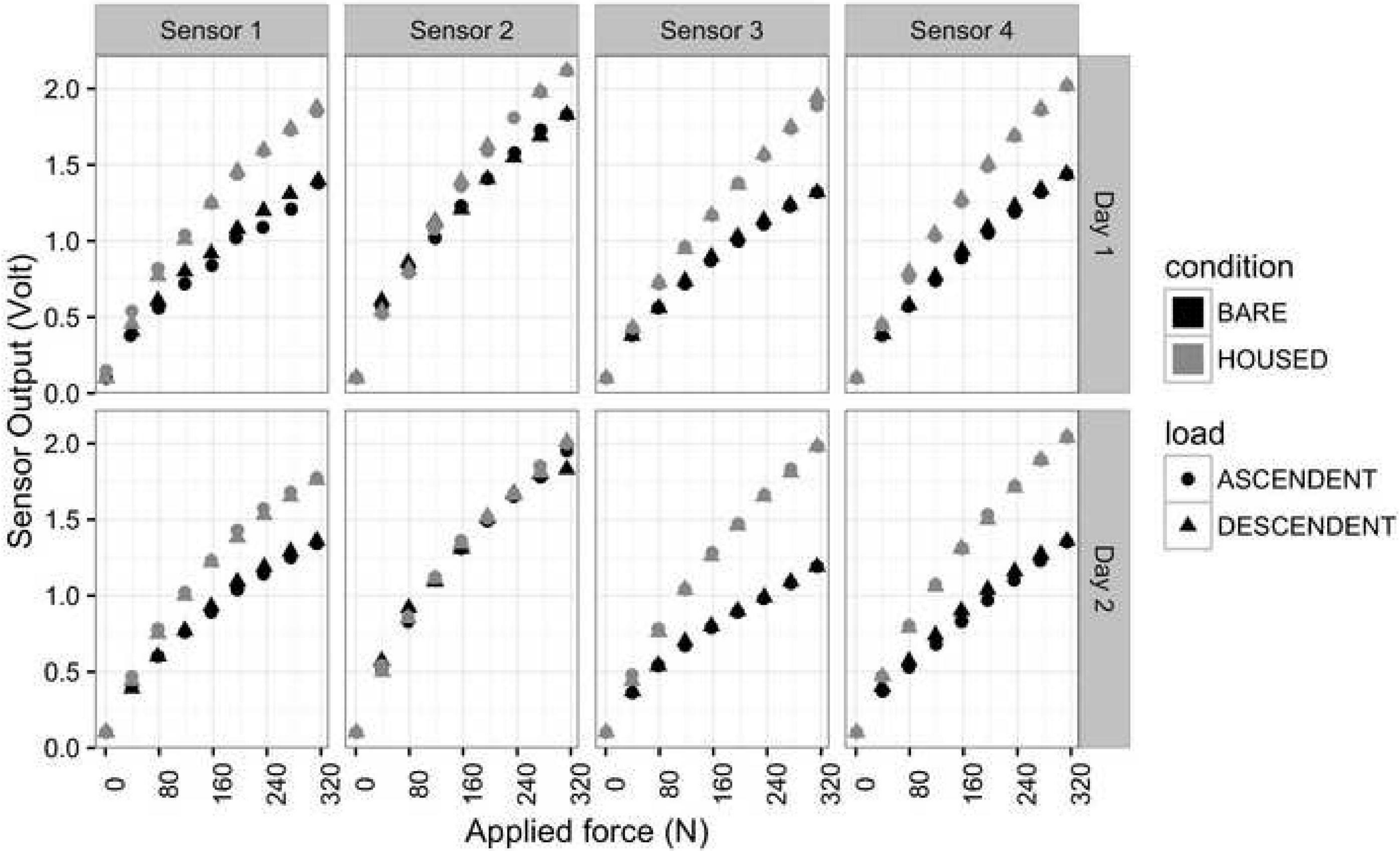

Figure 3B
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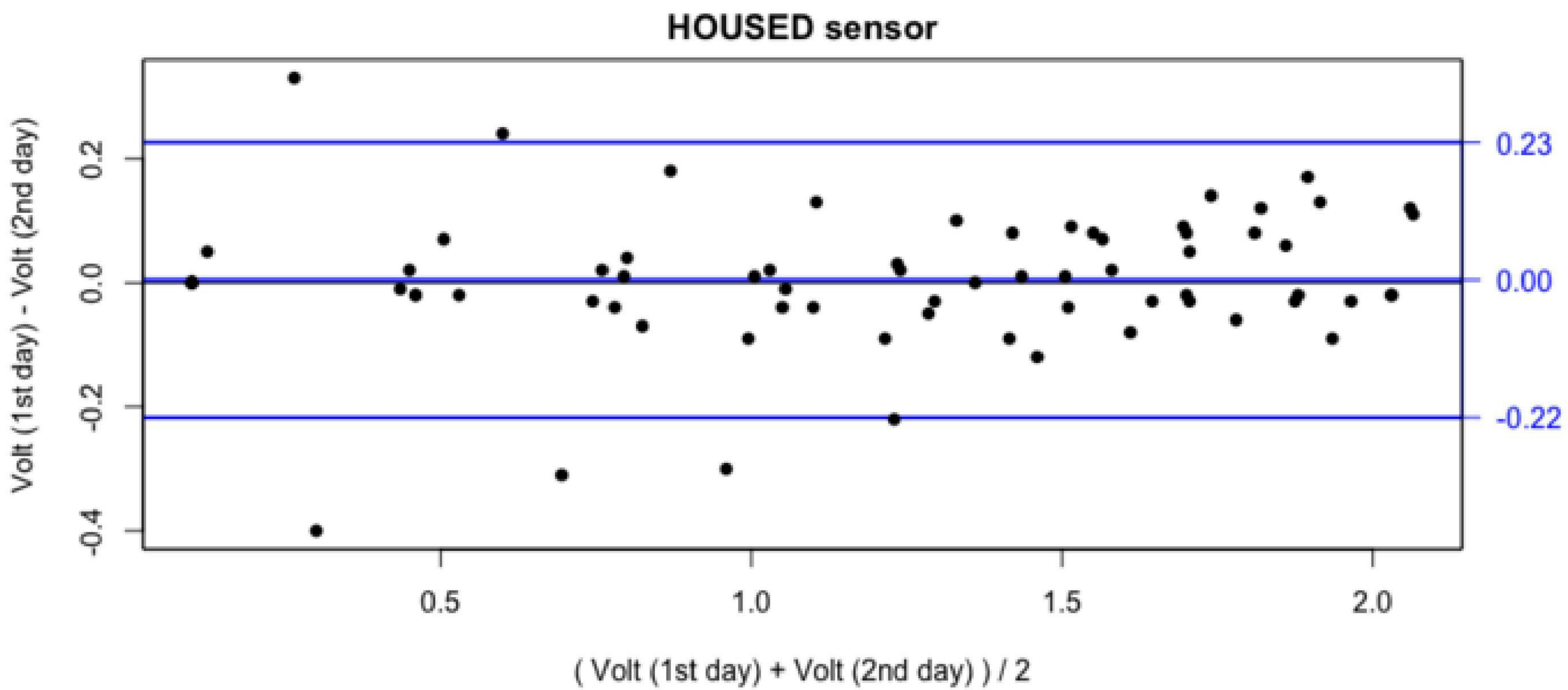

Figure 1:

A - *The piezoresistive film sensor*

B - *Housed transducer, top view*

C - *Housed transducer, lateral view: 1) external silicone layer, 2) internal hard plastic layer, 3) metal disk*

Figure 2: *Individual load-response curve of the four sensors in bare and housed conditions. Ascending (circle) and descending (triangle) for the bare (black) and housed sensor (grey).*

Figure 3: *Limit of Agreement of the four sensors: A) in bare condition; B) in housed condition.*

Tab 1: *Values of intercept (in V) and slope (in V/N) of each sensors in bare and housed conditions and in two days.*

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